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3D Dosimetric Tools in Radiotherapy for Photon Beams

Mirko Salomón Alva-Sánchez and Thatiane Alves Pianoschi

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1. Introduction

The technological advances to develop or improve devices for radiotherapy have increased in recent years, leading to conformality in the absorbed-dose within the tumor volume and avoiding healthy tissue. For quality assurance of the treatment, measurement of complex dose distribution from the new devices needs to be verified in the treated volume and healthy structure surrounding it.

According to ICRU the prescribed and administered dose must be within -5 % and + 7 % [1]. Many dosimetric protocols for radiotherapy recommend use of an ionization chamber like reference dosimeter, however, dosimeters like TLD, film, diode and other tools can be used although they measure the absorbed-dose at a point or in two dimensions [2,3].

A clinical system-dosimetry that can measure the dose distribution in 3D with tissue-equivalent characteristics is not available. In this context, dosimetry based on gel has shown a great potential for measured dose-distribution in 3D [4-7].

The MAGIC dosimeter with formaldehyde has characteristics of tissue-equivalence to water, stability of response and high detection resolution for suitable application in 3D dosimetry for complex treatments. Especially for use in techniques of high dose rate it can verify the volume dose as a quality control in radiotherapy. The gel dosimeter with formaldehyde (MAGIC-*f*) has been used in simulation studies as well as in clinical [8-11].

The most common used simulation algorithms, based on Monte Carlo calculations, are ITS2, EGS4, MCNP4 and PENELOPE [12-16]. The algorithm for mixed transport of charged particles implemented by PENELOPE algorithm led to its use in radiotherapy, simulating several irradiations and geometry of clinical situations in radiotherapy, which have shown a good concordance with experimental measurements using different irradiation techniques [17-20].

This chapter will approach two dosimetry tools: the MAGIC-*f* gel dosimeter and the PENELOPE-Monte code. Also, this chapter will cover some clinical applications of the two dosimetry tools with modalities like brachytheray, radiosurgery and 3D conformal therapy.

2. MAGIC-*f* dosimeter

Gel-based dosimeters called polymer gel dosimeters have been proposed since 1954 [21-24]. In 1992, formulations like BANANA, BANG and PAG polymer gels were proposed [25,26]. The suppression of polymerization by oxygen, present in these gels, was improved with a proposition of MAGIC gel (Methacrylic and Ascorbic acid in Gelatin Initiated by Copper) [27], which allows the MAGIC polymeric gels to be prepared in normal weather conditions [28-30], which for higher temperatures beyond 25 °C loss stability. Thus, was add formaldehyde to the original formulation of the MAGIC, re-calling as MAGIC-*f*, which led to a rise in its melting point to 69 °C [31].

The monomeric compounds of the MAGIC-*f* dosimeter immersed in an aqueous matrix, after exposure to irradiation suffer a polymer reaction, resulting in a polymer gel matrix. This formation changes the NMR relaxation properties of the gel and can be related to the deposited energy. One way to show this change in 3D is by using magnetic resonance images.

3. PENELOPE code

PENELOPE (Penetration and Energy Loss of Electrons and Positron) is a package used to simulate the transport of electrons, positrons and photons in arbitrary materials and complex geometries. The package has written on a FORTRAN platform, which include characteristics of many materials, and a user file. Besides these files, the code has a database with the characteristics of various materials of interest in radiological physics [32], cross section libraries and other quantities necessary for the transport of particles [33]. The simulation algorithm is based on a model that combines numerical and analytical cross sections for different types of interactions and is applied to initial energies from 1 keV to 1 GeV. The code has been used in some applications to simulate different irradiation techniques, reproduce the dimensions of irradiation geometry, source, energy, distance and other dosimetric parameters.

4. Application of the two dosimetric tools for some modalities in radiotherapy

4.1. Water-equivalent calibration of ^{192}Ir HDR brachytheray source using MAGIC-*f* gel

HDR brachytherapy is a treatment technique that uses sealed radioactive sources to treat prostate, colorectal cancer and some gynecological malignancies [34,35]. The commonly used implants are of ^{192}Ir sources with a half-life of 74 days, emitting beta rays ranging from 530 keV to 670 keV, and a gamma ray with an energy of 370 keV [36].

Due to the energy spectrum present in this source and the high gradient dose, a proper source calibration is necessary, although vendors assign large uncertainties to the calibration values (up to ± 10%). The calibration protocols for HDR brachytherapy sources recommend a specification in reference to air kerma rate using ionization chamber [37,38]. Absorbed dose measurement in water has recently been proposed to calibrate ^{192}Ir HDR brachytherapy sources [39].

For determination of the water-equivalent calibration of an ^{192}Ir HDR source we used the MAGI-f dosimeter, which has been shown to be a suitable dosimeter for many treatments in radiotherapy due to its characteristics of water equivalence (effective atomic number of 7.41), spatial resolution better than 1mm and the most important characteristic of measuring dose tridimensionally. Through the MAGIC-f were determined dosimetric parameters like percentage dose depth (PDD), and calibration curve (CC).

The two dosimetric parameters, PDD and CC, were obtained with the three dosimetric tools a)thermoluminescent dosimeter, b)ionization chamber and c)PENELOPE-Monte Carlo code simulation. The results obtained with these dosimetric tools were compared with the obtained data of polymer gel [40].

4.1.1. Materials and Methods

A water phantom of 50 x 50 x 50 cm^3 with electronic positioning holder (0.1mm precision) and the Gamma Med Plus 232 ^{192}Ir source with 5.43 Ci (200.91 GBq) was employed in this work. This source has an active cylindrical volume of 0.6 mm diameter and 3.5 mm height encapsulated in stainless steel welded to a steel cable, as shown in figure 1.

Measurements for PDD determinations, in water, were carried out using TLDs (LiF-100 with dimensions: 0.9 x 0.9 x 3.1 mm^3) packed with plastic and a plane parallel ionization chamber (Markus type, 0.05cm^3) after proper calibration. Both dosimeters were introduced in different depths (from 3 up to 12 mm) in a water phantom (figure 2). The catheter containing the HDR source was set parallel to the TLDs and the ionization chamber.
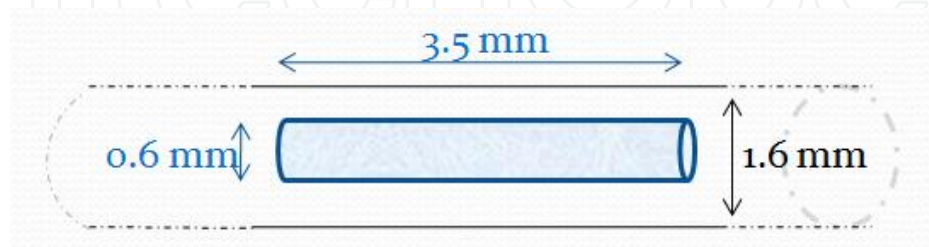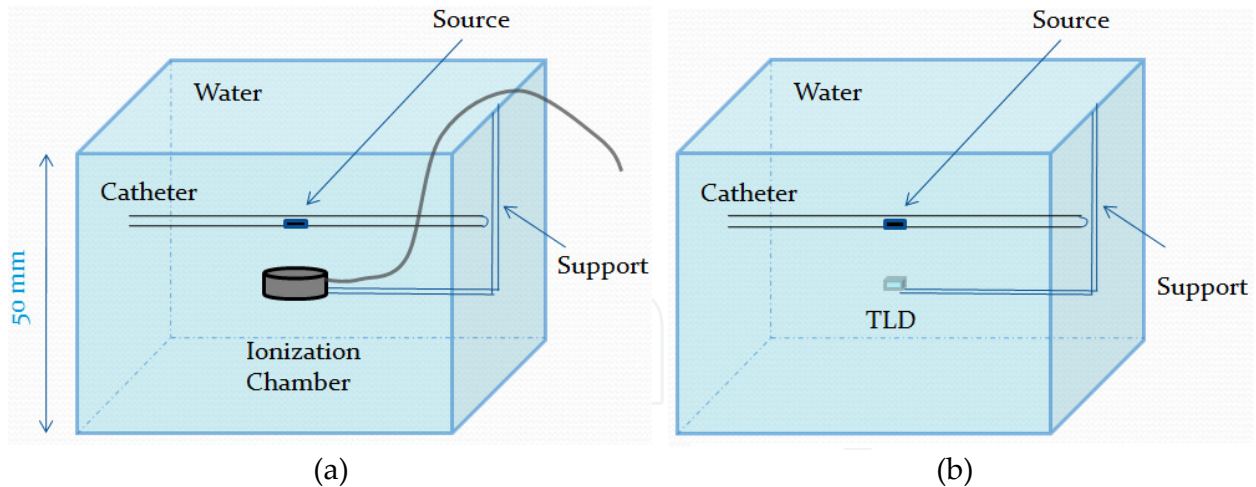

Figure 1. Design of ^{192}Ir source

Figure 2. Measurements setup for PDD determination with (a) ionization chamber and (b) TLDs.

Magic gel procedures

After preparation the solution of gel, this was poured in an acrylic cylindrical phantom of 6 cm diameter, 8 cm height and 0.5 cm thickness taking care to prevent air bubbles(figure 3).

During the PDD measurement the phantom filled with the gel was kept inside the water phantom and the source was positioned at a distance of 2 cm from the gel phantom wall.
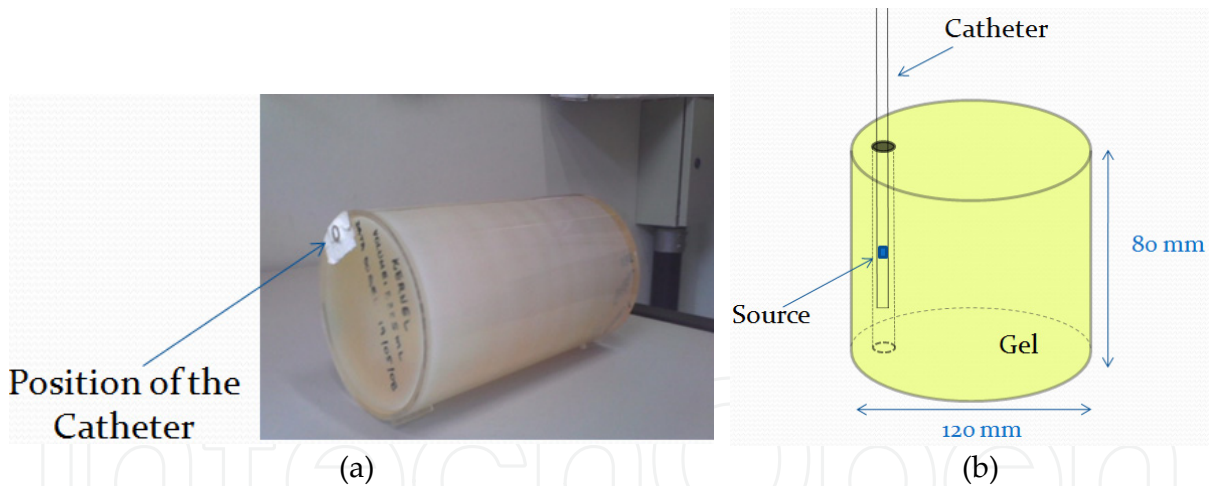

Figure 3. Cylindrical phantom to obtain the PDD curve: (a) with the MAGIC-f gel and (b) scheme of the irradiation setup.

For gel calibration, part of the gel was poured into three cylindrical glass tubes with 5ml volume, 12mm diameter and 75 mm height, closed with a 20mm hermetic stopper inserted in a plastic cover, as shown in figure 4.

Relaxation images of the gel phantom and glass tubes were acquired using a 1.5 T scanner (Siemens, Magneton Vision) one day after the irradiation to allow enough time for gel reaction completion. A head coil and multi spin echo sequences with 16 echo times in multiples of 22.5 ms, a repetition time of 3000 ms and a matrix size of 512 × 512 pixels were used during image acquisition. The slice thickness was 2mm. The transverse relaxation rate $R2$ (=$1/T2$) was

calculated by fitting the signal intensities versus the echo time pixel by pixel. The R2 maps were related to absorbed dose using a specific program developed in MatLab® 6.5.
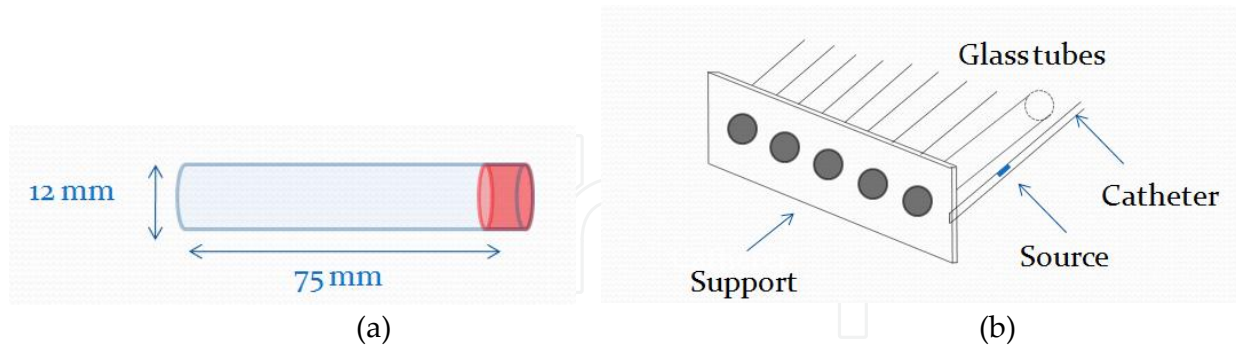

Figure 4. (a) Glass tube of 5 ml and (b) scheme of the setup to determine the calibration curve for the gel dosimeter.

PENELOPE-Monte Carlo

The user code was written to describe the geometry of the source and simulate all anisotropic properties of the brachytherapy extended source. A cubic water phantom were simulated, with dimensions of 30 cm, appropriate to obtain full scattering conditions. Simulation spatial resolution was set to1 mm.

4.1.2. Results and conclusions

Calibration curves for HDR [192] Ir source using TLD, ionization chamber and MAGIC gel are shown in figure 5.

The three dosimeters showed linearity with relative dose with correlation coefficients (R) of 0.9948, 0.9959 and 0.9971 for CI, TLD and the gel respectively. Maximum uncertainty of 3.5 % was found for measurements with the TLD for 10 Gy and a maximum uncertainty of 0.8 % was found for 10 Gy using the other dosimeter. The CC curve obtained with the TLDs did not show sensitivity for low doses (0.5 Gy) and MAGIC gel presented a background dose, approximately of 1.3 s^{-1}, which was subtracted for PDD measurements.

The calibration curve obtained with the Magic-f gel for the HDR [192] Ir source shows that it is possible to calibrate this source in water as stipulated in the protocol for external source [41]. Since, the gel is constituted by 90% water, effective-atomic-number of 7.41, which can be represent interaction of radiation with the water.

For PDD determinations, dose map distributions were obtained with MAGIC-f in a few planes as shown in figure 6.

The PDD curves were determined through percentage absorbed dose in depth, D_d, relative to the absorbed dose in the maximum depth, D_m, as show in equation 1

$$PDD(d) = \frac{D_d}{D_m} \, x \, 100\% \tag{1}$$
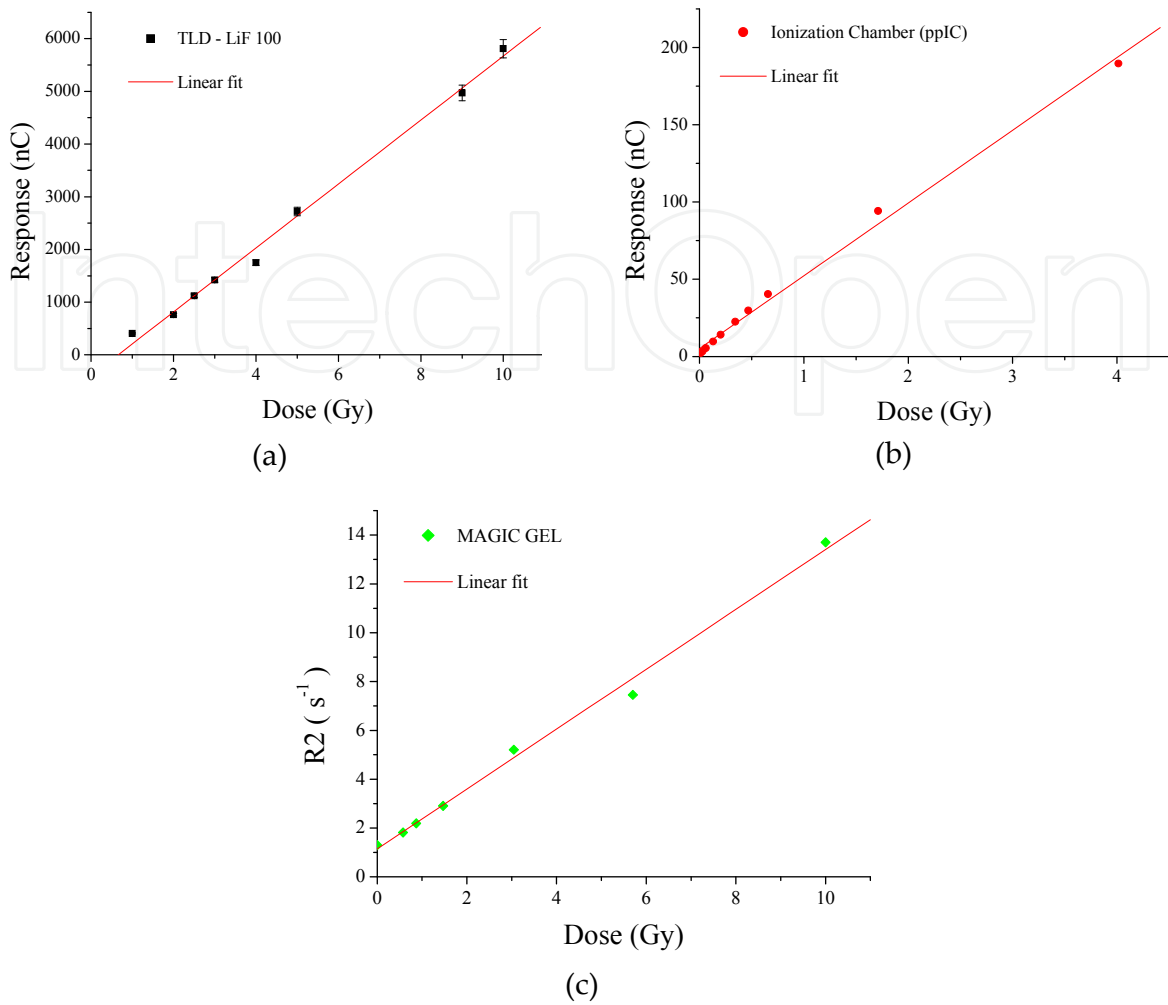
(a)

(b)

(c)

Figure 5. HDR [192] Ir source calibration curves: (a) TLD LiF-100 , (b) IC and (c) MAGIC polymer gel

The dose maps were obtained through the relaxation images and analysed with the program, in MatLab® developed to relate the nuclear resonance signal of the image with the absorbed dose.
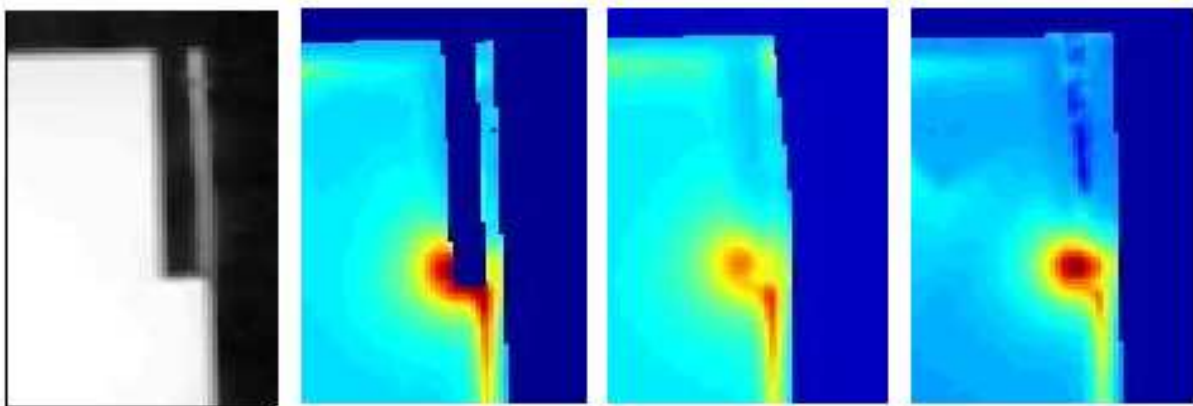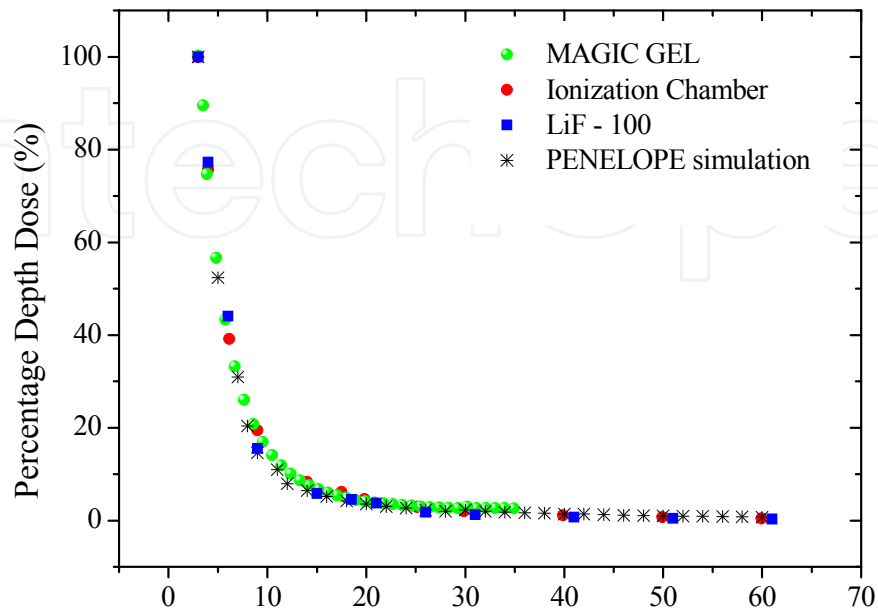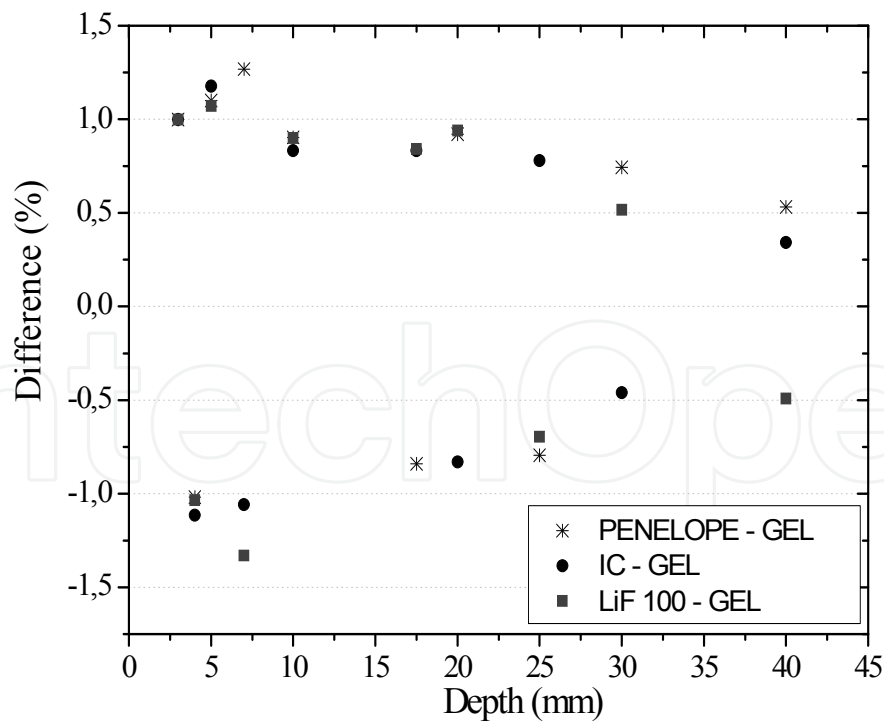

Figure 6. Relaxometry image and dose maps around the HDR [192] Ir source obtained experimentally with MAGIC polymer gel.

The PDD curves obtained by TLD, ionization chamber, MAGIC gel and PENELOPE code are shown in figure 7.

(a)

(b)

Figure 7. Percentage depth dose for HDR ¹⁹²Ir with MAGIC polymer gel, ionization chamber, TLD LiF-100 and PENELOPE code, (b) differences between PENELOPE, IC and TLD-LiFs related to MAGIC gel.

From the percentage depth dose and calibration curve results it can be concluded that the MAGIC polymeric gel can be used as dosimeter in routinely clinical procedures and can be used to check the calibration of [192]I sources in an water-equivalent medium as part of the assurance quality in HDR brachytherapy.

4.2. Dose-distribution of small fields through the MAGIC-*f* dosimeter and PENELOPE code

Some techniques in radiotherapy, like intensity-modulated radiation and stereotactic radiosurgery, use small radiation fields and higher dose-radiation. Since the dosimetry of the small fields is more critical than that for large fields, due to no-equilibrium conditions created as consequence of the secondary electron track length and the real source size project from the collimator to the phantom or patient. Other complication is the perturbation of the level of disequilibrium, which is inducted bye the dosimeter when introduced in the track of the radiation fields. Thus, due these characteristics of the small fields, an accurate dose verification of the delivery dose of the small radiation fields is required [42-45].

The aim of this application is evaluate the response of MAGIC-f dosimeter, after being exposed to a conformal irradiation technique using a small field to compare the results with PENELOPE code through two dimensional dosimetric parameters [46].

4.2.1. Materials and Methods

Treatment planning system (TPS)

The TPS BrainSCAN 5.31, was used to plan a conformal irradiation with the tomography images of the cylindrical phantom, with 9 irradiation fields from 0° to 340°. The prescribed dose was determined through the calculated algorithm using pencilbeam for 16 Gy in 100% of the target volume.

MAGIC-f gel and PENELOPE code

The solution of Magic-f gel was poured in a cylindrical phantom of PPMA with dimensions of 12 cm diameter, 18 cm height and thickness 0.5 cm. The phantom containing the gel was placed at source-phantom center distance of 100 cm, figure 8, from the linear accelerator (Siemens MEVATROM with multileaf MX2), irradiated with 9 fields of 6 MV photon beam, each one of 1 x 1 cm², to a total dose of 16 Gy. The experimental irradiation was made in the same irradiation conditions as the TPS.

PENELOPE simulations were used to determine the percentage depth dose, beam profile and a conformational dose distribution in the same irradiation and geometry condition made by the MAGIC-f.

After the irradiation, the reading was performed through MRI using a 3 T scanner and head coil. Relaxation images were obtained using a multi spin echo sequence with 16 echo times in multiples of 22.5 ms, a repetition time of 3000 ms and a matrix size of 256 × 256 pixels. The slice thickness was 2mm for each slice.

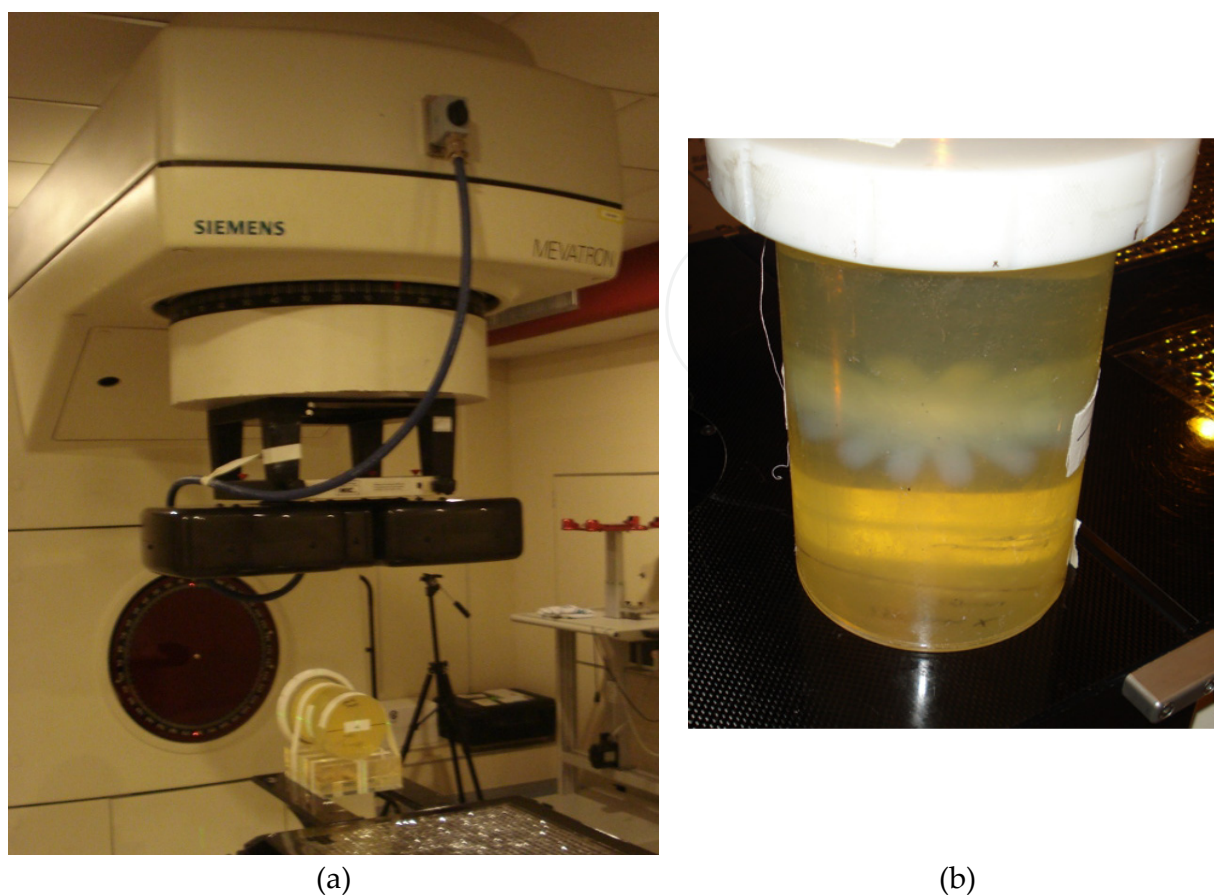(a)   (b)

Figure 8. (a) Setup of the irradiation with the linear accelerator coupled with the multileaf, (b) cylindrical phantom after irradiation with 9 fields

PENELOPE simulations were used to match the irradiation and geometry conditions as the experimental irradiation. The spatial resolution for the simulations was 0.5 mm.

The two dose distributions obtained with the gel and PENELOPE were compared through beam profiles for each irradiated field.

4.2.2. Results and conclusions

The two dosimetric tools were validated through comparison PDD and BP curves. The determination of the PDD was made through the equation 1. The PDD curves are shown in the figure 9 for an irradiation field of 1 x 1 cm^2, source-phantom center distance of 100 cm with energy of 6 MV and absorbed dose of 5 Gy.

The PDD curves obtained with the two dosimetric tools show the same behavior. Maximum differences of 6.5 % were found in 10.1 cm depth comparing the two curves.

The BP determinations were made in the line along of maximum dose depth (perpendicular to the beam irradiation) through the equation 2, since D_n are de dose in the line where is the maximum dose, D_m. The BP determinations were irradiated with the same irradiation of the PDDs. Both BPs curves are show in figure 10.

$$BP(\%) = \frac{D_n}{D_m} \, x \, 100\%$$

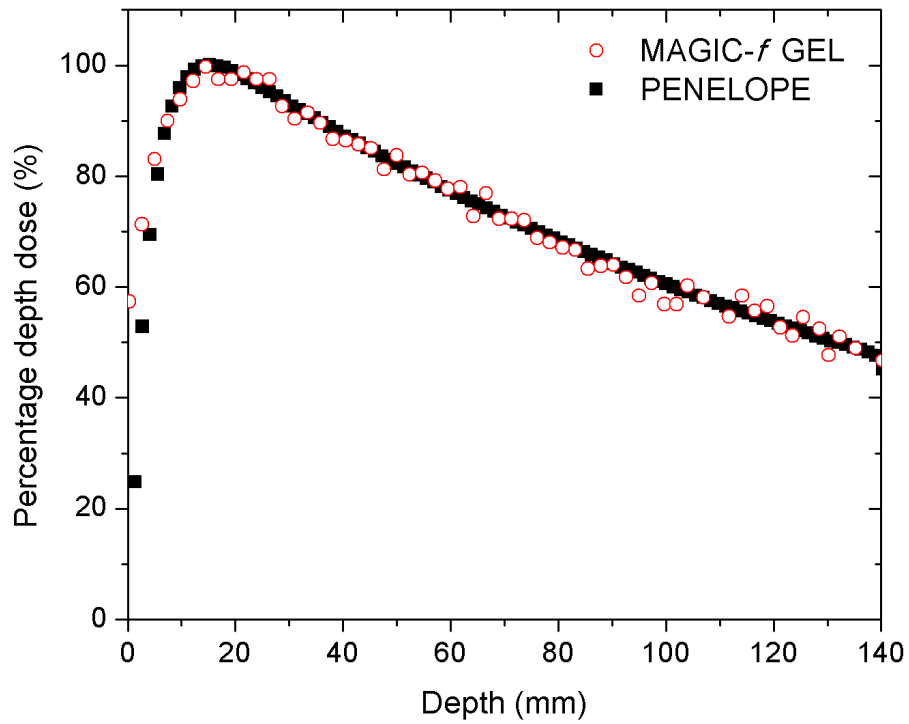<div align="right">(2)</div>

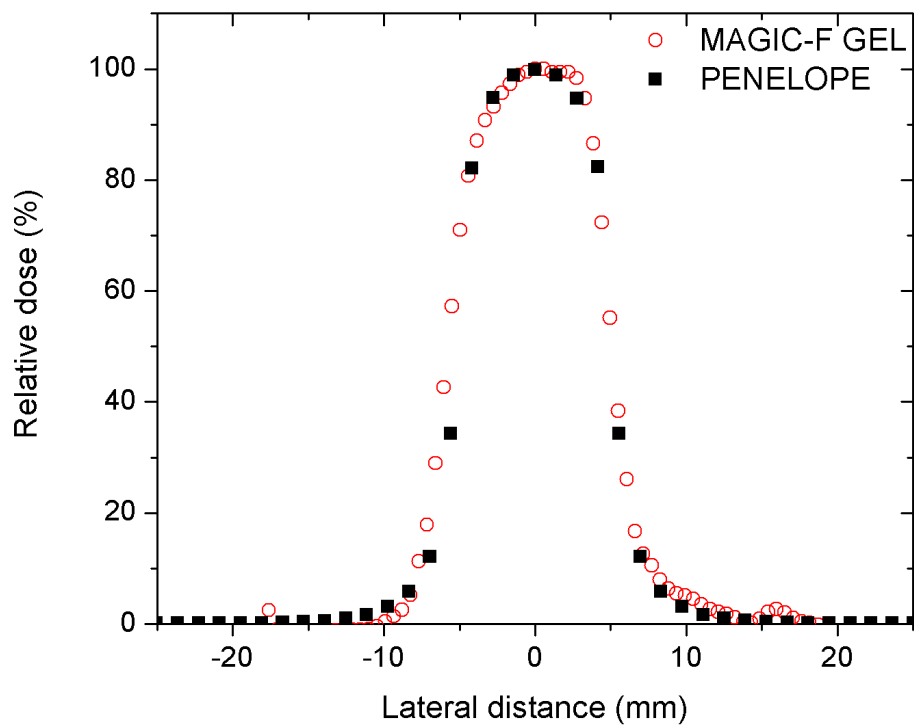**Figure 9.** Percentage depth doses obtained with the MAGIC-f gel and PENELOPE code for a irradiation field of 1 x 1 cm²

Figure 10. Beam profile obtained with the MAGIC-f gel and PENELOPE code for 1 x 1 cm².

The BP curves obtained with the two dosimetric tools show the same behavior, with maximum differences of 4.3 %, where compare both curves in distances more than size of the irradiation field. Inside the irradiation field (1 cm²) the maximum difference was of 0.9 %.

The dose-distributions to a conformal irradiation technique obtained with tool dosimeters, gel and PENELOPE, are showed in the figure 11. The dose-distributions obtained were compared through the BP, as show in figure 12.
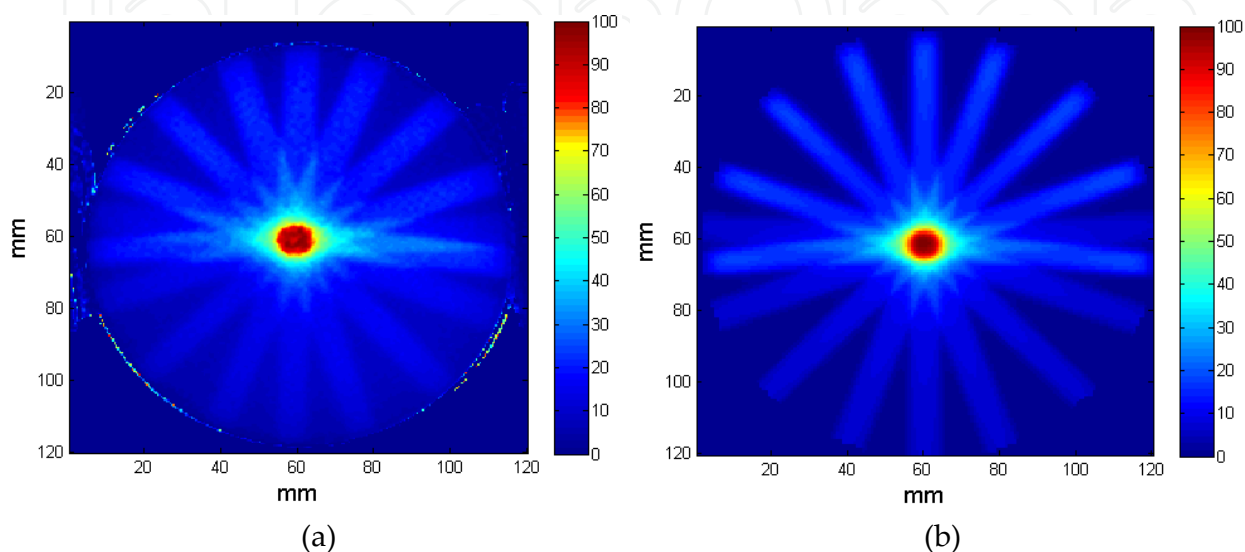

(a)   (b)

Figure 11. Dose-distributions for a conformal irradiation: (a) MAGIC-f dosimeter and (b) PENELOPE code

The comparisons between gel and PENELOPE showed maximum differences of 0.91% inside the isodose of 50% .For isodose major of 50% the difference was up to 12.3%.
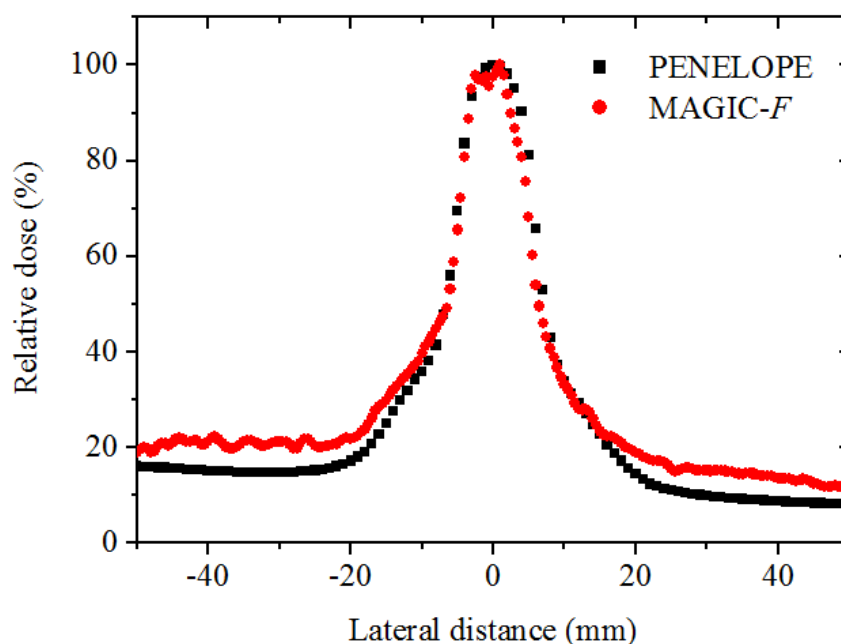

Figure 12. Beam profiles obtained with the PENELOPE code and the Magic-f gel.

From the results obtained it can be infer that the response of the MAGIC-*f* polymeric gel is suitable and can be used routinely as a dosimeter in clinical procedures using small fields, especially in radiosurgery.

4.3. Gamma index comparison for the conformal dose-distribution trough dosimetric tools

The modern radiotherapy imposes high level of accuracy to deliver absorbed-dose in a complex way, which depends on beam energy, field size, geometry of the irradiation target and others aspects. Thus, the generated dose-distributions must be checked rigorously, both for administered and delivered dose distributions.

MAGIC-f gel with high spatial resolution, effective-atomic-number close to water and an ability to measure the dose in 3D and also, PENELOPE code are available with a high concordance and estimate measurement of dose distribution in 3D.

Comparisons of dose-distributions obtained with the simulation and experimental measures can be analyzed through a beam profile, as showed in the section 4.2, which give us a limited information of that comparison. A subtraction pixel by pixel of the dose map technique can offer more information but is not related the tolerance of the position. The gamma index is a technique that relates to tolerance with the dose and position within a limited range [47,48]. Dose-distributions obtained with the MAGIC-f gel and PENELOPE simulation were compared through the technique of gamma index [49].

4.3.1. Materials and methods

The dose-distributions obtained with the MAGIC-f gel and PENELOPE code were compared by determining gamma index (γ) test. A program was developed in MalLab®, which calculated the gamma index; following the parameters of distance-to-agreement (DTA) and the tolerance of dose difference. The DTA determines the comparison tolerance of localization of one pixel from the MAGIC-f dose-distribution to the pixels within circumference, for an established radius, from the PENELOPE dose-distribution. Likewise, the dose tolerance parameter determines the comparison tolerance of dose value of one pixel from the MAGIC-f dose-distribution to the dose values to pixels within circumference, for an established radius, from the PENELOPE dose distribution.

Thus, the formulation for determining the γ values, combining the two criteria: DTA and dose tolerance, are expressed in the equation 3:

$$\Gamma = \sqrt{\frac{r^2(r_m, r)}{\Delta d_M^2} + \frac{\delta(r_m, r)}{\Delta D_M^2}} \tag{3}$$

Since:

$r(r_m, r) = |r - r_m|$ is the difference between the position of a reference point of PENELOPE dose distribution, r_m, with the, r, point of the Magic-f dose-distribution, that is being compared.

$\delta(r_m, r) = D(r) - D_m(r_m)$ is difference of doses in the position r and r_m.

$\Delta d_M^2 e \Delta D_M^2$ are the criteria of DTA and dose difference, respectively.

Thus, to quantify the quality of the γ index for any point within range of r-r_m is chosen the minimum value of that range, which express by equation 4.

$$\gamma(rm) = \min\{\Gamma(rm, r)\} \forall \{r\} \tag{4}$$

the γ test indicate that the compared of dose distribution can be consider similar into of the criteria of the DTA and dose difference or not:

$\gamma(r_m) \leq 1$, indicate that the comparison is acceptable

$\gamma(r_m) > 1$, indicate that the comparison not is acceptable

4.3.2. Results and conclusions

As shown in the section 4.4, we obtained dose-distributions for the conformational irradiation of 9 fields of 1x1 cm², using the Magic-f gel and the PENELOPE code.

Both dose-distributions were compared using the gamma index test through the program developed using the criteria of DTA of 3 mm and dose tolerance of 3 %. Figure 13 show the γ map, with maximum values up to 2.5.
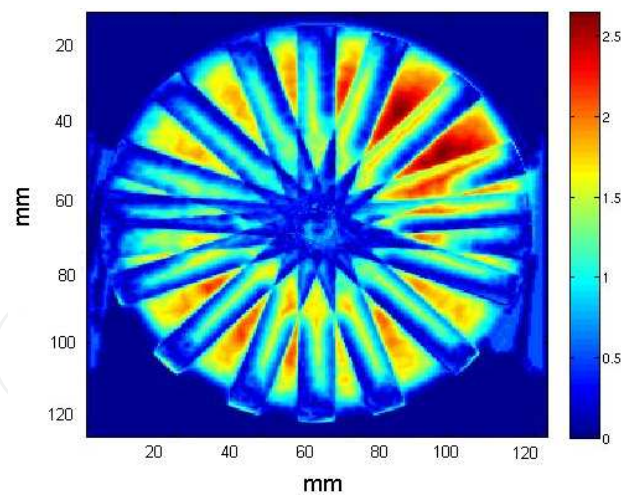

Figure 13. Gamma index value, obtained from comparing dose-distribution between Magic-f and PENELOPE.

Analyzing a central region, target volume, of 1 cm diameter from the dose-distribution obtained by the gel and PENELOPE code, 100% of the dose values pass the test under gamma index conditions. Although for regions farther than 1 cm only 57% of the dose values pass this test, which present gamma index values more than 1.

The results have shown that the response of MAGIC-f gel has a concordance with the simulated values in the central region. To improve the gel response some studies will be done to minimize the difference in region with high dose gradient and low doses to that hereafter it can be used like a dosimeter routinely in clinical procedures.

4.4. Monte Carlo simulation of MAGIC-*f* gel for radiotherapy using PENELOPE

A way to study and predict MAGIC-*f* gel results is the use of computational simulation as the Monte Carlo method. From the codes more commonly used in radiotherapy is the PENELOPE code. This code allows the "construction" of materials trough the use of the compound's chemical composition (*i.e.*, elements present and stoichiometric index, or weight fraction, of each element), mass density, mean excitation energy and energy and oscillator strength.

The aim of this application was use PENELOPE to simulate MAGIC-*f* gel dosimetric properties and to provide a valuable enhancement in the pre-constructed list of materials from PENELOPE. For validation the simulated material file (*MAGIC-f.mat*) the percentage depth doses (PDD) curves and a dose distribution were used, comparing experimental values obtained with MAGIC-*f* gel dosimeter and simulated results [50].

4.4.1. Materials and methods

The components of MAGIG-*f* gel (water, gelatin, methacrilic acid, copper sulfate, ascorbic acid and formaldehyde) and its characteristics, like atomic number, density and molar mass, were input into the PENELOPE 2008 code to build the *MAGIC-f.mat* and *water.mat* material files and for the simulation of PDDs and the conformal treatment. The simulations used 2×10^9 primary particles and $0.01 mm^2$ pixel size.

For experimental measurements were used MAGIC-*f* dosimeter. Relaxometry, weightened in T2, with a 3.0 T, NMR Philips tomography and head coil were used for the readings of the gel samples. The images were acquired with a multi spin-echo sequence with 16 echo times, TE=20ms, TR=4000ms, matrix of 256 x 256 pixels, slices of 3mm thickness. The MR images were processed and analysed with a program developed in MatLab®, that produces R2 maps.

PDDs and dose-distribution

Simulated PDDs were used to evaluate the *MAGIC-f.mat* constructed by the code. Simulation conditions were $10 \times 10 cm^2$ field size and 100 cm source-skin distance (SSD) for photon beams of 6 MV and 10 MV and phantoms of $30 \times 30 \times 20 cm^3$ "filled" with both simulated materials.

Experimental PDDs were determined using MAGIC-*f* gel and ionization chamber (IC) of 0.6 cc for 6 MV and 10 MV beams. The measurements with the gel were performed using glass tubes of 16 cm length and 1 cm diameter filled with gel. The irradiations were carried out using PPMA cube filled with water and the glass tubes positioned in the center of the volume. A dose of 10 Gy in conventional irradiation conditions for both beams was used.

For the measurement of a coplanar dose-distribution, the gel was poured into a PPMA cylindrical phantom of 10 cm diameter and 15 cm height. The irradiation of the phantom was made with 5 fields of 1 x 1 cm², with a Varian 2100 linear accelerator gantry angles of 0° to 180°. The target was chosen in the center of the cylinder, at 100 cm SSD. The phantom was irradiated with 10 MV beam with2 Gy per field. The conformal irradiation was simulated using PENELOPE code in the same irradiation conditions. Both distributions were also compared through the dose profile.

4.4.2. Results and discussions

PDDs and dose-distribution

Figure 14 shows the PDDs through the simulations and from experimental measurements, showing them similar behavior. A maximum difference of 1.93% and 1.88% was found for the 6 MV and 10 MV beams, respectively, when simulated PDDs with *MAGIC-f.mat* and *water.mat* are compared.
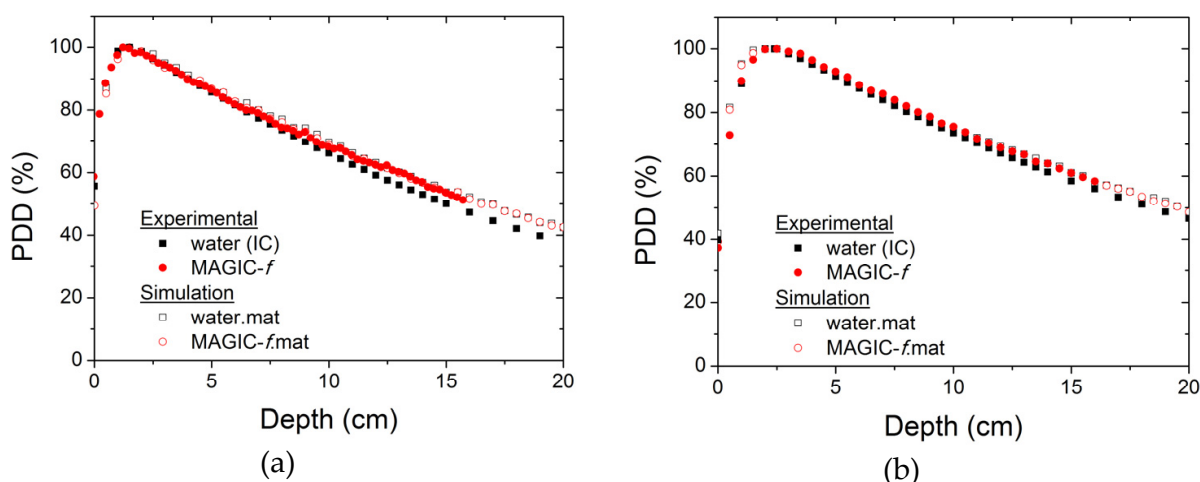

Figure 14. PDDs for water and MAGIC-*f* simulated with PENELOPE and experimental values for (a) 6 MV beam and (b) 10 MV beam.

For the 6 MV beam, a comparison between experimental gel and water (IC) shows a maximum difference of 3.20% until 10 cm depth. Beyond 10 cm the maximum difference is 5.72% in 12.5 cm. The same comparison was performed for the 10 MV beam and a maximum difference of 1.93% was found until 10 cm depth and of 2.62% at 14cm depth.

The maximum differences between experimental and simulated values are 1.46% (0.5 cm depth) and 4.3% (17cm depth) for 6 MV beam. Maximum differences of 2.14% (1 cm depth) and 3.8% (18cm depth) were found for the 10 MV beam.

The study of conformal dose distributions for 10 MV using 5 fields was performed through simulation using *MAGIC-f.mat*, shown in the figure 15.a. and experimentally using MAGIC-*f* gel dosimeter as shown in the figure 15.b.

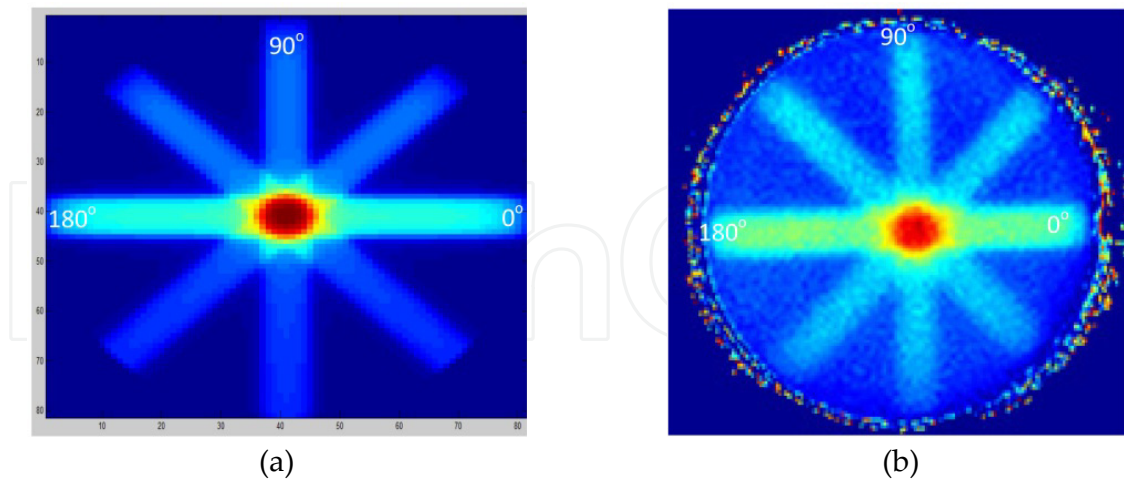Figure 16 shows the dose profiles obtained from both distributions.

Figure 15. Dose distribution for 10 MV obtained with: (a) PENELOPE; (b) MAGIC-f dosimeter.
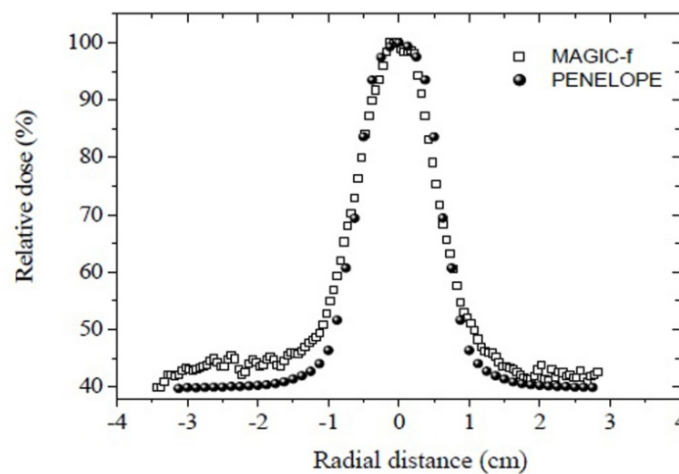

Figure 16. Dose profiles of simulated and experimental distributions.

A maximum difference of 0.5% in a radial distance of 0.55 cm inside the field (approximately half the irradiation field) was found and beyond that distance the differences between the dose distributions are less than 7%.

The simulations using PENELOPE code showed similar behaviors with the experimental values. Although discrepancies were found, specifically in the dose-distribution in regions of low doses, this study shows that the code can be used to simulate the components of the MAGIC-f gel for these energies. More over, the *MAGIC-f.mat* file can be useful for studies of response and comparison of the gel.

5. Conclusions

The validation of the MAGIC-f gel and PENELOPE through the dosimetric parameters, percentage depth dose and beam profile showed that that both dosimetric tools present similar behavior.

The use of two dosimeters for all radiotherapy modalities present in this work showed great concordance, although discrepancies were found in experimental measurements, using the MAGIC-f gel, and the simulations obtained with the PENELOPE code.

From the results obtained it can be infer that the response of the MAGIC-*f* polymeric gel and simulation by PENELOPE-Monte Carlo code are suitable and can be used routinely as a dosimeter, in case of the gel, and in the way to study or predict the response and shape of dose-distribution by the simulation.

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Acknowledgement

6. References

[1] International Commission radiation and units, ICRU, (1999) Prescribing Recording and Reporting Photon Beam Therapy (Supplement to ICRU Report 50), Report n° 62, ICRU, Bethesda, 53 p.

[2] American Association of Physics in Medicine, AAPM, (1994) Code of practice for radiotherapy accelerator: Report of AAPM radiotherapy task Grop no 45. Med. Phys. 21:1093-1121.

[3] International atomic energy agency, IAEA, (2000). ''Absorbed dose determination in external beam radiotherapy: An international code of practice for dosimetry based on standards of absorbed dose to water '', thechnical Report Series. IAEA TRS 398, IAEA, Vienna, Austira.

[4] De Deene Y (2002) Gel dosimetry for the dose verification of intensity-modulated radiotherapy treatments Z. Med. phys. 12: 77–88.

[5] De Deene Y, Hurley C, Venning A, Vergote K, Mather M, Healy B, Baldock C (2002) A basic study of some normoxic polymer gel dosimeters. Phys. med. biol. 47: 3441–3463.

[6] McJury M, Oldham M, Cosgrove V P, Murphy P S, Doran S, Leach M, Webb S (2000) Radiation dosimetry using polymer gels: methods and applications. Br. J. Radiol. 73: 919–929.

[7] Haraldsson P, Karlsson A, Weislander E, Gustavsson H, Bäck S (2006) Dose response evaluation of a low-density normoxic polymer gel dosimetry uising MRI. Phys. med. biol. 51: 919-928.

[8] Pavoni J, Pike T, Snow J, DeWerd L, Baffa O (2010) Dosimetria tridimensional usando gel MAGIC com formaldeído. Rev. bra. de física médica. 4: 15-18.

[9] Zhu X, Fakhri G, Reese T, Crowley E (2010) Improved MAGIC gel for higher sensitivity and elemental tissue equivalent 3D dosimetry. Med. phys. 37: 183-189.

[10] Pianoschi T, Alva M, Santanna M, Baffa O, Nicolucci P (2010) Magic-f gel dosimeter for clinical electron beam. J. phys. conf. ser. 250: 012037.

[11] Schwarcke M, Marques T, Garrido C, Nicolucci P, Baffa O (2010) MAGIC-f gel in Nuclear Medicine Dosimetry: study in an external beam of Iodine- 131. J. phys. conf. ser. 250: 012082.

[12] De Vlamynck K, Palmans H, Verhaegen F (1999) Dose measurements compared with Monte Carlo simulations of narrow 6 MV multifead collimator shaped photon beams. Med. phys. 26: 1874-1882 .

[13] Yamamotro T, Mizowaki T, Miyabe (2007) An integrated Monte Carlo dosimetric verification system for radiotherapy treatment planning. Phys. med. bio. 52: 1991-2008.

[14] Al-Ghorabie F, Natto S, Al-Lyhiani S (2001) A comparison between EGS4 and MNCP computer modeling of an in vivo X-ray fluorescence system. Comp. bio. med. 31: 73-83.

[15] Chiavassa S, Lemosquet, Aubineau-Lani L, de Carlan L, Clairand I, Ferrer L, Bardi M, Franck D, Zankl M (2005) Dosimetric comparison of Monte Carlo codes (EGS4, MCNP, MCNPX) considering external and internal exposures of the zubal phantom to electron and photon source. Radiat. protec. Dosim. 116: 631–635.

[16] Neto A, Haddad C, Pelosi E, Zevallos-Chávez J, Yoriyaz H, élio Yoriyaz, Siqueira P (2005) Monte Carlo Simulation as an Auxiliary Tool for Electron Beam Quality Specification for Intra-Operative Radiotherapy. Braz. J. phys. 35: 801.

[17] Sempau J, Badal A, Brualla L (2011) A PENELOPE-based system for the automated Monte Carlo simulation of clinacs and voxelized geometries—application to far-from-axis fields. Med. phys. 38: 5887-5895.

[18] Koivunoroa H, Siiskonen T, Kotiluoto P, Auterinen I, Hippelainen E, Savolainen S (2012) Accuracy of the electron transport in MCNP5 and its suitability for ionization chamber response simulations: A comparison with the EGSNRC and PENELOPE codes. Med. phys. 39:1335-1344.

[19] Ramirez J, Chen F, Nicolucci P, Baffa O (2011) Dosimetry of small radiation field in inhomogeneous medium using alanine/EPR minidosimeters and PENELOPE Monte Carlo simulation. Radiat. Measur.46:941-944.

[20] Gonzales D, Requena S, Williams S (2012) Au La x-rays induced by photons from 241Am: Comparison of experimental results and the predictions of PENELOPE. Appl. Radiat. Isot. 70: 301–304.

[21] Alexander P, Charlesby A, Ross M (1954) The degradation of solid polymethylmethacrylate by ionization radiation. Proc. r. soc. A: 223-392.

[22] Boni A (1961) Polyacrylamide gamma dosimeter. Radiat. Res. 14: 374-380.

[23] Audet C, Schreiner L (1991) Radiation dosimetry by NMR relaxation time measurement of irradiated polymer solutions. Proc. Soc. magn. reson med. 10th Annual Scientific Meeting 705.

[24] Maryanski M, Gore J, Kennan R, Schulz R (1993) NMR relaxation enhancement in gels polymerized and cross-linked by ionizing radiation: a new approach to 3D dosimeter by MRI. Mag. reson. imaging. 11: 253-258.

[25] Maryanski M, Schultz R, Ibbott G, Gatenby J, Xie J, Horton D, Gore J (1994) Magnetic resonance imaging of radiation dose distributions using a polymer-gel dosimeter. Phys. med. biol. 39: 1437-1455.

[26] Baldock C, Burford R, Billinghan N, Wagner G, Patval S, Badawi R, Keevil S (1998) Experimental procedure for the manufacture of polycrylamide gel (PAG) for magnetic resonance imaging (MRI) radiation dosimetry. Phys. med. biol. 43: 695-702.

[27] Fong P, Keil D, Does M, Gore J (2001) Polymer gels for magnetic resonance imaging of radiation dose distributions at normal room atmosphere. Phys. med. biol. 46: 3105-3113.

[28] Baldock C, Burford R, Billinghan N, Wagner G, Patval S, Badawi R, Keevil S (1998) Experimental procedure for the manufacture of polycrylamide gel (PAG) for magnetic resonance imaging (MRI) radiation dosimetry. Phys. med. biol. 43: 695-702.

[29] De Deene Y, Hurley C, Venning A, Vergote K, Mather M, Healy B, Baldock C (2002) A basic study of some normoxic polymer gel dosimeter. Phys. med. biol. 47: 3441-3463.

[30] Gustavsson H, Karlsson A, Back S, Olsson L, Haraldsson P, Engstrom P, Nystrom H (2003) MAGIC-type gel for three-dimensional dosimetry: Intensity-modulated radiation therapy verification. Med. phys. 30:1264-1271.

[31] Fernandes J, Pastorello B, de Araujo D, Baffa O (2008) Formaldehyde increases magic gel dosimeter melting point and sensitivity. Phys. med. biol. 53: N1-N6.

[32] INTERNATIONAL COMIMISSION ON RADIATION UNITS AND MEASUREMENTS, ICRU, (1989)Tissue Substitutes in Radiation Dosimetry and Measurement. ICRU Report 44, Washington.

[33] Salvat F, Fernandez-Varea J, Acosta E, Sempau J (2005) PENELOPE – A Code System for Monte Carlo Simulation of Electron and Photon Transport, Nuclear Energy Agency OECD/NEA, Issy-les-Moulineaux, France. Available: http://www.nea.fr. Accessed 2011 Nov 02.

[34] Yan X, Poon E, Reniers B, Voung T, Verhaegen F (2008) Comparison of dose calculation algorithms for colorectal cancer brachytherapy treatment with a shielde applicator. Med. phys. 35: 4824-4830.

[35] Takam R, Bezak E, Yeoh E (2009) Risk of secondary primary cancer following prostate cancer radiotherapy: DVH analysis using the competitive risk model. Phys. med. biol. 54: 611-625.

[36] Goggen T (1988) Physical Aspects of Brachytherapy. Medical Physics Handbooks editors, England, pp 198.

[37] Nath R, Anderson L, Meli J. Olch A, Stitt J, Williamsom J (1997) Code of practice for brachytherapy physics: Report of the AAPM Radiation Therapy Committee Task Group No. 56. Med. phys. 24:1557-1598.

[38] International Atomic Energy Agency, IAEA, (2002) Calibration of photon and beta ray sources used in brachytherapy. IAEA-TECDOC 1274.

[39] Sarfehnia A, Stewart K, Seuntjens J (2007) An absorbed dose to water standard for HDR ^{192}Ir brachytherapy sources based on water calorimetry: numerical and experimental proof-of-principle. Med. phys. 34: 4957-4961.

[40] Alva M, Marques T, Schwarcke M, Gonçalvez L, Baffa O, Nicolucci P (2009) Water-equivalent calibration of ^{192}Ir HDR Brachytherapy source using MAGIC-f gel Polymer. In: World Congress on Medical Physics and Biomedical Engineerin. IFMBE Proceedings. Munich : Springer Berlin Heidelberg. 25: 248-251.

[41] AGÊNCIA INTENACIONAL DE ENERGIA ATOMICA, IAEA, (2005) "Determinación de la dosis absorbida en Radioterapia con haces externos: un código de práctica internacional par la dosimetria basada en patrones de dosis absorbida en agua". Viena: Agencia Internacional de Energia Atômica, IAEA-informe técnico 398.

[42] Babic S, Battista J, Jordan K (2009) Three-dimensional dosimetry of small megavoltage radiation fields using radiochromic gels and optical CT scanning. Phys. med. biol. 54: 2463-2481.

[43] Wong C, Ackerly T, He C, Patterson W, Powell C, Qiao G, Solomon D, Meder R, Geso M (2009) Small field size dose-profile measurements using gel dosimeters, gafchromic films and micro-thermoluminescent dosimeters. Radiat. mea. 44: 249-256.

[44] Calcina C, Oliveira L, Almeida C, Almeida A (2007) Dosimetric parameters for small field sizes using Fricke xylenol gel, thermoluminescent and film dosimeters, and an ionization chamber. Phys. med. biol. 52: 1431-1439.

[45] Das J, Ding G, Ahnesjö A (2008) Small fields: Nonequilibrium radiation dosimetry. Med. phys. 35: 206-215.

[46] Alva M, Pianoschi T, Takeda F, Alves T, Haddad C, Nicolucci P. (2010) Dose Distribution of Small Fields through MAGIC-F Gel Dosimetry and PENELOPE-Monte Carlo Simulation. Med. phys. 37: 3122.

[47] Low D, Harms W, Mutic S, Purdy J (1998) A technique for the quantitative evaluation of dose distributions. Med. phys. 25:656-661.

[48] Low D, Dempsey J (2003) "Evaluation of the gamma dose distribution comparison method. Med. phys. 9:2455–2464.

[49] Alva M, Pianoschi, Amaral L, Oliveira H, Nicolucci P(2011) Gamma Index comparison for the conformational dose distribution obtained through three dosimetric tools. Braz. J. Med. phys. 5: 66.

[50] Alva M, Pianoschi T, Marques T, Santanna M, Baffa O, Nicolucci P (2010) Monte Carlo Simulation of MAGIC- gel for Radiotherapy using PENELOPE. J. phys. conf. ser. 250: 012067.